Cardiac Imaging: 1.5T vs. 3.0T - Where's the Benefit?

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Introduction

The task of high resolution imaging of an organ in constant motion has led cardiovascular magnetic resonance (CMR) technology to continually push the speed limits, and thus the signal-to-noise (SNR) limits, of MRI. Since SNR is directly proportional to static magnetic field strength (B_0), CMR theoretically stands to benefit from higher field magnet systems. A 1998 revision of USFDA guidelines [1] allowed MRI systems up to 4.0T to qualify as non-significant risk devices. Since then, all major manufacturers have commercially released whole-body 3.0T MRI systems for clinical use. Cardiac array coils with 8-channels or more have also been available for 3.0T MRI for some time, permitting the use of parallel acquisition techniques for increased imaging speed. A number of centers have explored the 3.0T potential of doubled SNR for cardiovascular imaging. Some CMR applications at 3.0T have lived up to the promise of increased SNR, speed, and/or resolution, while others have not. This presentation will review some of the theoretical advantages and practical challenges of CMR at 3.0T, and provide a summary of current research on this topic.

Theoretical Benefits of 3.0T vs. 1.5T

The primary motivation for the development and manufacture of whole-body 3 Tesla MRI systems is the theoretical increase in the intrinsic signal-to-noise (ISNR) [2] which comes with higher field. ISNR is purely determined by the electrodynamics independent of relaxation, homogeneity and other effects which may reduce observable SNR gains [3]. Several recent comparisons of pulse sequences at 1.5T and 3.0T have demonstrated close to the theoretical 2X increase in SNR in cardiac imaging applications [4-9]. The observed gain in SNR is influenced by a number of factors, including receiver coil design, B_0 and B_1 field homogeneity, and RF flip angle limitations governed by increased RF power deposition (SAR). Hence, the observed SNR gain is typically somewhat less than linear with field strength.

In many CMR applications, increased speed and/or resolution are desired, and SNR can be traded for these by the common relationships governing MR image characteristics. For example, SNR is linearly proportional to each voxel dimension, so higher field strength can be used to increase resolution while maintaining sufficient SNR. As another example, increasing the sampling bandwidth (BW) results in a reduction in minimum TR, and therefore faster data acquisition, but increased BW also causes a direct loss of SNR by the factor $\sqrt{BW}$. This strategy of faster scanning by increased receiver BW faces two practical limits at 3.0T. First, as TR is decreased, SAR increases. Second, gradient amplitude and slew rate limits ultimately restrict the resolution obtainable for a given BW. Higher performance gradient sets would be useful to take advantage of the higher SNR at 3.0T, but commercially available gradient systems at 1.5T and 3.0T are already at the limits determined by physiological stimulation. Parallel acquisition techniques [10-13] provide an alternative means of trading SNR for acquisition speed independent of gradient system performance.

Longitudinal relaxation time (T1) increases with field strength. The recent study by Stanisz et al, [14] showed myocardium T1 increased by 43% from 1030 msec at 1.5T to 1471 msec at 3T, and blood T1 increased 34% from 1441 msec at 1.5T to 1932 msec at 3T. In other tissues, increases in T1 from 1.5T to 3T ranged from 10% in cartilage to 73% for kidney. T1 lengthening can be both an advantage and a disadvantage in CMR, depending on the imaging application. Saturation tags last longer at higher field due to slower T1 recovery[7]. Spin labeling techniques benefit from the longer T1 times at 3T and may be feasible for quantification of myocardial

perfusion in the human heart [15]. The relaxivity of paramagnetic contrast agents, such as the common Gd-based agents, is only slightly reduced at 3T [16]. Thus, the increase in baseline tissue T1 at 3T can be an advantage when T1-shortening paramagnetic contrast agents are used. This combination leads to a potentially larger signal difference between contrast-enhanced tissue or blood, and un-enhanced tissue. This property can be exploited in contrast-enhanced MR angiography at 3.0T [16-18], as well as myocardial perfusion [5] and viability imaging [19]. While T2 relaxation of tissues at 3T is relatively unchanged from 1.5T [14], T2* is shorter due to microscopic magnetic field inhomogeneities increasing linearly with field strength. Susceptibility induced dephasing due to substances such as deoxyhemoglobin in the blood, or hemosiderin in the liver, and paramagnetic contrast agents are proportional to applied field. This increased sensitivity to susceptibility increases the sensitivity to the BOLD effect in the heart, and to the T2* shortening caused by iron particle based contrast agents. Current developments in iron-oxide particle based plaque-targeting and stem cell labeling will be more sensitive at higher field strength.

Finally, cardiac spectroscopy [20] benefits not only from more signal per unit volume per unit time at 3T, but also from the increased frequency dispersion between spectral lines.

Limitations of CMR at 3.0T

Unfortunately, the list of limitations and disadvantages of imaging at 3T is just as long as the list of advantages. The Larmor (resonant or precessional) frequency increases linearly with applied field strength. Specific absorption rate (SAR) depends on the square of frequency, since more energy is transmitted and absorbed at higher frequency for a given amplitude RF signal. The absorbed RF energy for equivalent pulse sequence parameters will increase by roughly a factor of four when field strength is doubled from 1.5T to 3.0T. Several CMR applications like SSFP cine, contrast-enhanced MRA, and black-blood turbo spin echo can already be SAR limited at 1.5T. This limitation can be severe at 3.0T, and leads to compromises in flip angle and TR which can reduce contrast and SNR for these techniques.

As mentioned in the list of advantages, local field distortion caused by differences in tissue susceptibility increases in proportion to field strength. This can be especially problematic in the heart where large differences in susceptibility are caused by the lung-tissue interface. Techniques such as SSFP and hybrid GRE-EPI are particularly sensitive to local field distortions and achieving equivalent image quality to that obtained at 1.5T with these techniques has proven difficult. Additionally, shorter T2* reduces signal in gradient echo imaging in general.

Significant RF (B_1) inhomogeneity and flip angle variation can occur within the body at 3T. The RF wavelength gets shorter at higher field [21]. Proton wavelength at 3T is approximately 26cm (52 cm at 1.5T), smaller than the typical dimensions of the torso, and can lead to destructive interference within the human body. This may be less of an effect in chest than abdomen due to the gaseous lungs, but spatial variation of flip angle can be problematic at 3T when trying to observe regional variation in tissue signal caused by pathology [8].

Increased T1 at 3T can be a disadvantage, depending on the application, due to increased saturation effects. Inversion times (TI) for blood or selective tissue nulling are longer, and may adversely effect the timing of sequence events within the cardiac cycle. Gains in SNR in may be compromised by slower T1 recovery between RF excitations.

Fat suppression performance is similar at 1.5T and 3.0T due to the counter-balancing effects of field distortion and chemical shift frequency difference between water and fat; both scale linearly with field strength. A similar counterbalancing effect is evident in spectroscopy; the frequency dispersion between spectral lines increased, but local field inhomogeneity increases linearly with field strength.

Obtaining a reliable ECG-signal for cardiac synchronization can be problematic at higher field. Blood flow in an applied magnetic field gives rise to induced voltages in the aorta and other major arteries that can be observed as superimposed electrical signals in the ECG. These voltages are dependent on field strength [22]. The safety of imaging patients at 3T with implanted devices such as stents [23] and pacemakers [24] must also be examined carefully at 3T.

CMR Applications at 3.0T

Black-blood

Spin echo based acquisitions are relatively insensitive to B_0 inhomogeneity, but the technique can be sensitive to RF inhomogeneity, and SAR limitations can be a problem in the case of single-shot acquisitions. Greenman et al [8] performed a comparison of black-blood fast spin echo sequences in normal volunteers at 1.5T and 3.0T. While double-inversion blood suppression performed comparably at both field strengths, they found a significant spatial variation in signal intensity at 3.0T attributable to B1 inhomogeneity. Tissue characterization capability may be compromised by any underlying spatial variation in signal intensity. Black-blood vessel wall imaging in carotid and coronary arteries [4, 25, 26] has been successfully demonstrated at 3T. This application pushes the resolution and SNR limits of MRI, and stands to benefit greatly from higher field strength. However, optimized array coils for high resolution carotid imaging are not yet commercially available at 3T. Plaque characterization by targeted iron oxide particle contrast agents is also expected to benefit from the higher SNR and T2* sensitivity at 3T.

Cine and flow

The SSFP sequence with its high SNR and blood-myocardium contrast has become the standard method of cine imaging at 1.5T [27]. Cine is arguably the most important component of any CMR exam, and unfortunately the SSFP technique is one of the most sensitive to the negative effects of higher field strength. The short TR and high flip angles of SSFP lead to high SAR, and the characteristic SSFP frequency response produces dark bands and flow artifacts in the presence of field inhomogeneity. Despite these limitations, recent work on optimization of SSFP techniques for 3T has shown the feasibility of obtaining images of equivalent quality to 1.5T [7, 28, 29]. RF pulse optimization to reduce SAR, and further improvement of volume localized shimming methods are necessary for successful SSFP imaging at 3.0T. Spoiled gradient echo (FLASH) cine, on the other hand, does not suffer from the same sensitivity to SAR and field inhomogeneity. Functional imaging techniques based on FLASH, such as myocardial tagging [7] and phase-velocity mapping [6], have been successfully applied at 3T, demonstrating the advantage of the higher inherent SNR.

First-pass perfusion and delayed-enhancement

These techniques are primarily based on spoiled gradient echo acquisition, and rely on the differences in contrast agent concentration in tissues for image contrast. Initial first-pass perfusion results at 3T are just coming out in the literature [5] showing imaging performance advantages over 1.5T. However, no studies in ischemic patients have been published to date. Optimization of first-pass imaging techniques specifically for 3T continues [30, 31]. Myocardial viability imaging by delayed-enhancement at 3T has only recently been demonstrated in patients [32, 33], with one of the two studies [33] showing CNR advantages over 1.5T imaging.

Coronary MRA

The promise of higher SNR at 3.0T is especially appealing for coronary MRA which has struggled to meet the SNR and resolution requirements needed for routine clinical application. Coronary MRA techniques are generally based on a SSFP readout, and excellent fat suppression is essential, so field homogeneity is especially critical to its success. This fact has made implementation at 3T difficult, and while some promising results have been demonstrated [34-38] it is still in a stage of technical evolution

Summary

So, where is the benefit for CMR at 3.0T? Data from clinical studies is just starting to come out, but the expected gains in SNR appear to be realizable in contrast-enhanced methods like first-pass perfusion, delayed-enhancement, and MR angiography. In addition, gradient-echo based applications like tagging and velocity mapping are able to reap the direct benefits of higher field strength without some of the deleterious “side effects” of increased SAR and field inhomogeneity. Further optimization of shim and methods of reducing SAR are necessary before SSFP-based techniques can be as reliable at 3T as they have proven to be at 1.5T.

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